(12) **UK Patent Application** (19) **GB** (11) **2 058 511 A**

(54) **Positron annihilation imaging device using multiple offset rings of detectors**

(57) A positron annihilation imaging device comprises two or more coaxial circular arrays of detectors (2, 2'), with the detectors in one array angularly offset with respect to the detectors in the adjacent array to detect more than one tomographic image simultaneously through different cross-sections of a patient.

Fig I

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Fig 1

Fig 2

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Fig 4

Fig 3

SPECIFICATION
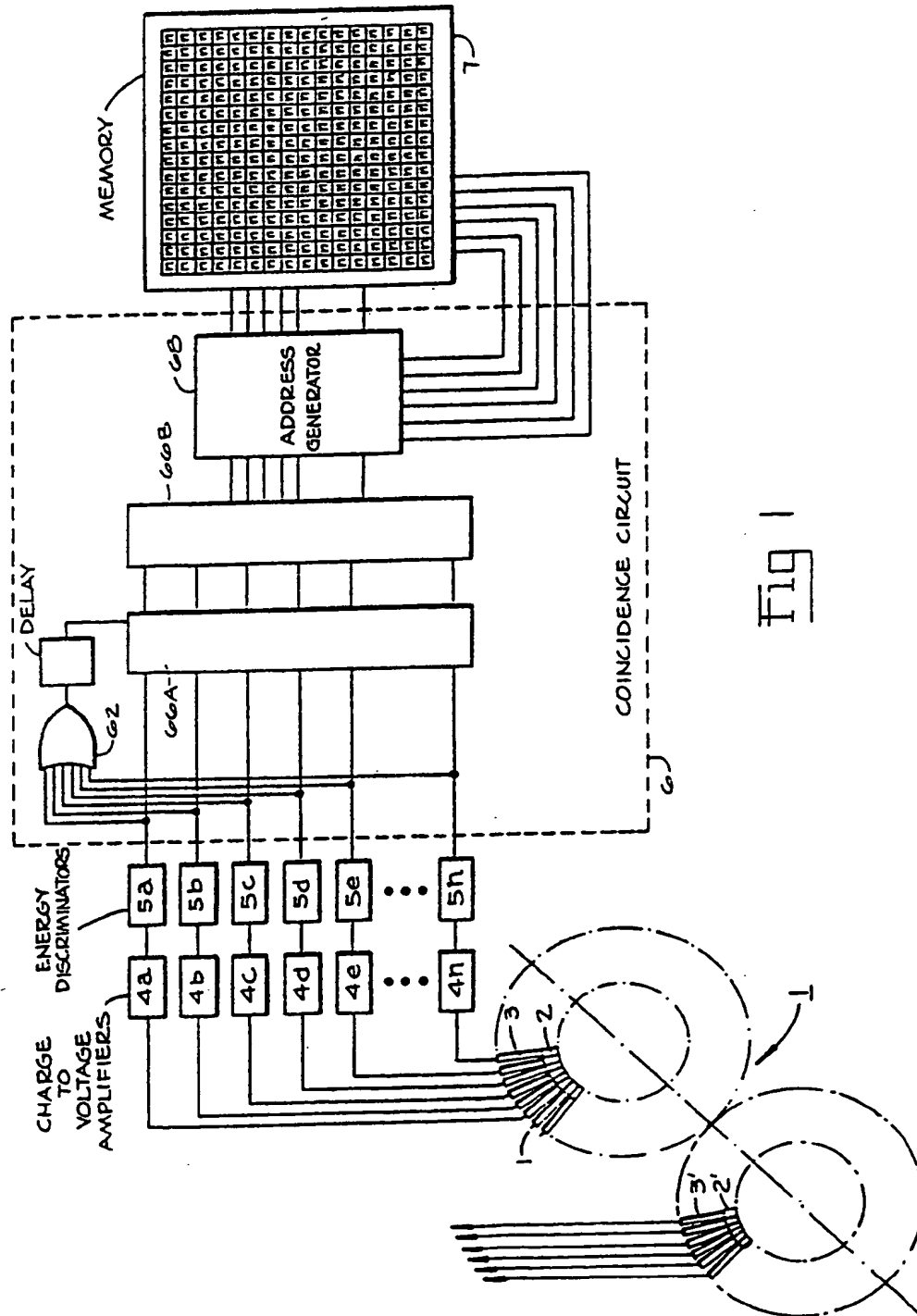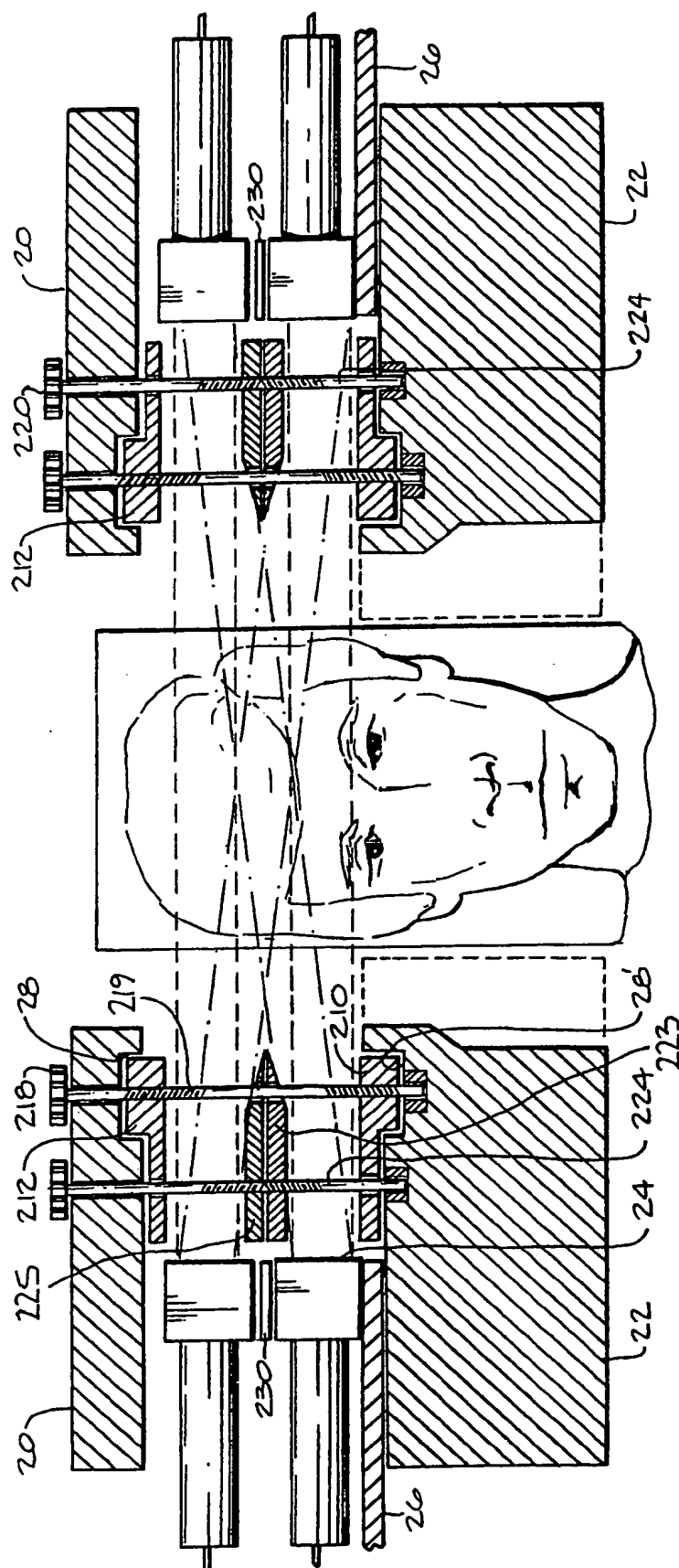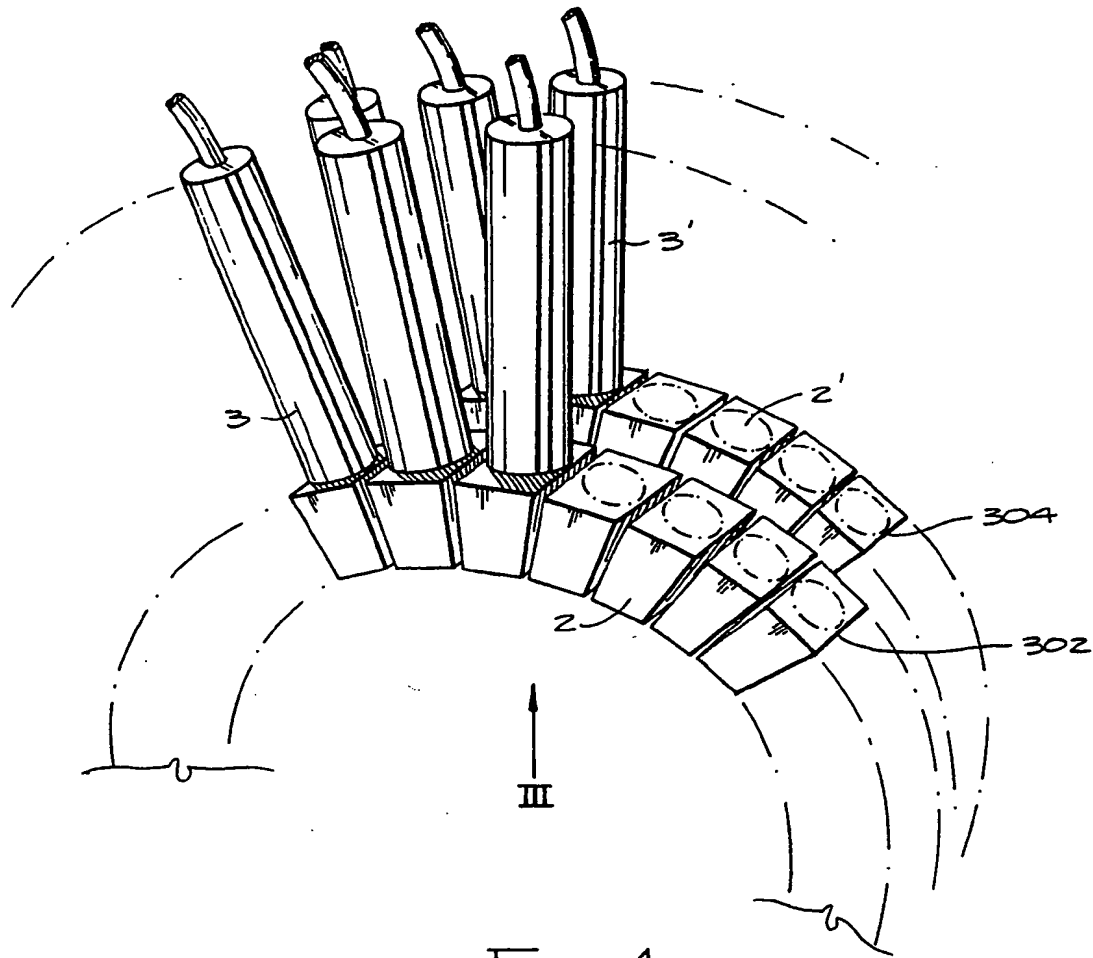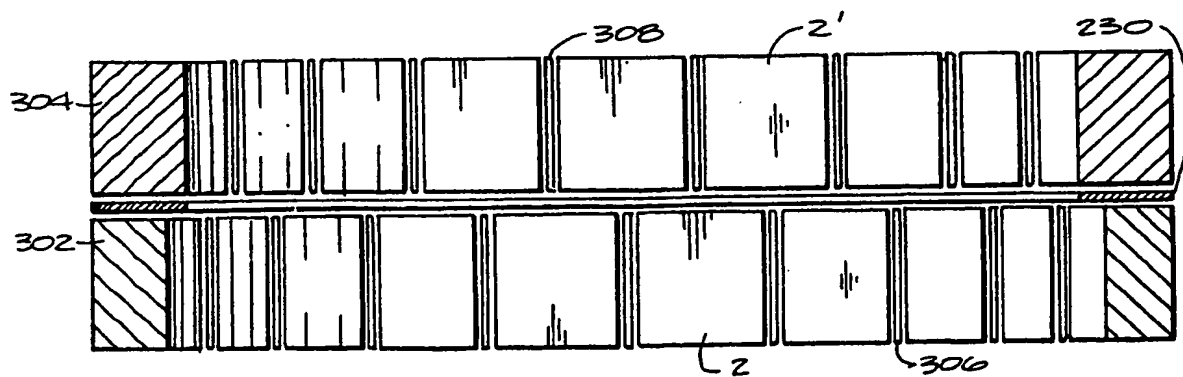**Positron annihilation imaging device using multiple offset rings of detectors**

This invention relates generally to positron
5 emission tomography and more particularly to
devices which use an array of scintillation
detectors to detect the annihilation radiation from
positron disintegration and use this information to
reconstruct an image of the distribution of
10 positron emitting isotopes within a body.
Positron emission tomography is a technique
for measuring the concentration of a positron
emitting isotope through a sectional plane through
the body. Normally the isotope is used to label a
15 substance which circulates with the blood and
may be absorbed in certain tissues. The technique
allows the actual concentration in the slice to be
determined if the device is suitably calibrated.
Certain isotopes decay by emitting a positively
20 charged particle with the same mass as the
electron (positron) and the neutrino from the
nucleus. In this process one of the protons in the
nucleus becomes a neutron, so that its atomic
number goes down while its atomic weight
25 remains constant. This positron is ejected with a
kinetic energy of up to 2MeV depending on the
istope and loses this energy by collisons while
travelling a distance of up to a few mms in water.
When it has reached thermal energies it interacts
30 with an electron and they mutually annihilate one
another. The rest mass of the 2 particles is
transformed into 2 gamma rays of 511 keV which
are emitted at 180° in the 'center of mass'
coordinates of the original particles. The 2 gamma
35 rays may be detected by suitable devices. If these
devices measure the energy of the gamma rays at
511 keV and register this energy almost
simultaneously it may be assumed that the origin
of the radiation is on a straight line between the 2
40 detectors. Several detectors may be used in an
arrangement so that many coincident events may
be imaged during the same time interval. Then the
information from these detectors is processed by a
computer using image reconstruction techniques
45 in order to find the location of distribution of the
positron emitting isotope.
A device for imaging positron annihilation
radiation comprises the following basic parts:
(1) A number of detectors arranged in a precise
50 geometrical pattern. These detectors are normally
scintillation detectors in one or several planes, and
these detectors are normally arranged in a
polygonal pattern or around the circumference of
a circle. Scintillation detectors emit a light flash
55 each time they absorb gamma radiation which
may or may not arise from the mutual annihilation
of a positron and electron. The intensity of the
light flash is proportional to the gamma ray
energy.
60 (2) The device must contain a means of
converting the light flash to an electrical charge
pulse. Its amplitude is proportional to the light
intensity.
(3) The device must contain a means of

65 determining that the charge pulse could have
arisen from a gamma ray whose energy was
approximately equivalent to the mass of the
electron at rest (511 keV).
(4) The device must have an electric circuit
70 capable of determining that 2 and only 2 detectors
each recorded gamma rays or appropriate energy
within a short time interval (coincidence resolving
time). These detectors are said to have recorded a
'coincident event'.
75 (5) The device must have an electric circuit
which determines which 2 detectors out of the
many possible combinations recorded the so-
called 'coincident event'.
(6) The device must have a memory in which it
80 can record how often each pair of detectors record
a 'coincident event'. The memory may be part of
the random access memory of a general purpose
computer.
(7) The device is required to use an algorithm
85 through which the information in the memory may
be transformed into an image of the distribution of
positron annihilation per unit time in a cross-
section surrounded by the detectors. The
sequence of steps described by this algorithm may
90 be programmed into a general purpose computer.
Accordingly it is a main object of this invention
to provide a means for recording more than one
tomographic image simultaneously through
different cross-sections of a patient.
95 In accordance with the present invention there
is provided a positron annihilation imaging device
comprising:
(a) a first array of N detectors equally spaced
around a first circle and disposed in a first
100 common plane; and
(b) a second array of N detectors equally
spaced around a second circle and disposed in a
second common plane, the first and second circles
being coaxial, said first and second planes being
105 parallel and spaced apart, the detectors in the first
array being offset angularly with respect to said
detectors in the second array by an angle
substantially equal to 360/2N° subtended at the
said common axis.
110 Some embodiments may also include at least
one additional array of N detectors equally spaced
around and axially circularly disposed in one or
more planes spaced from said common planes,
the detectors in said additional array, being offset
115 by 360/2N° relative to the detectors in an
adjacent one of said first or second arrays of
detectors.
The invention is described further hereinafter,
by way of example, with reference to the
120 accompanying drawings, in which:—
Fig. 1 is an overall block diagram of an
apparatus embodying the invention;
Fig. 2 is a cross-sectional view showing how
the coincident events are obtained from two rings
125 of detectors to produce three independent
tomographic images of consecutive cross-section
through a patient;
Fig. 3 shows the detector placement in a two-
ring system which is able to produce three cross-

sectional slices; and

Fig. 4 is a pictorial view of two rings of detectors, the detectors in one ring being offset in relation to the detectors in the outer ring.

5   A preferred embodiment of the positron annihilation imaging device 1 is shown in outline form in Fig. 1. The device comprises two or more rings of detectors 2, 2' which surround the object being imaged in two or more planes. The electrical
10   signals from these detectors are amplified at 4a ... 4n, and their energy is measured in energy discriminators 5a ... 5n, the outputs of the energy discriminators being processed by a coincidence circuit 6. The output of the coincident circuit is
15   used to increment memory locations in a general purpose computer. The computer then reconstructs an image of the distribution of the positron imaging isotope in the cross-sections which were scanned.
20   Further details of the circuit of Fig. 1 will be found in our copending U.K. patent application No. 8003837 entitled "Coincidence Analysis Circuit for Positron Annihilation Imaging Device".

The preferred embodiment contains two rings
25   of 64 trapezoidal shaped bismuth germanate detectors which are separated by thin tungsten septa. (See Fig.2).

For further details of the detector shape and arrangement for a positron annihilation imaging
30   device, reference may be made to our copending U.K. patent application No. 8003836 entitled "Scintillation detector for annihilation radiation in positron disintegration process".

These two rings of detectors are rotatable with
35   respect to one another by half the angular separation of the detectors (2.8°).

In normal operation of a single slice positron annihilation detector ring it is desirable to rotate the single detector ring in an oscillatory motion to
40   and fro by half the angular separation of the detectors. The purpose of this is to double the number of points in each parallel projection which is used in the image reconstruction technique. The other purpose of this is to uniformly sample the
45   object to be scanned in such a way that the sampling of the projection is done at points which are not greater than half the widths of the detector pair aperture function apart. This is necessary to prevent "aliasing errors" in the reconstructed
50   image.

This oscillatory motion takes a finite time (in the preferred embodiment about one third of a second) which limits the rate at which consecutive images can be obtained to approximately two
55   thirds of a second per image. An image can be reconstructed of a 'central slice' between the two planes of detectors when two rings of detectors are used as shown in Fig. 2. Since this central slice is viewed by twice the number of detectors, twice
60   the number of counts per unit time are recorded by the detectors from this area. Because of this, an image can be reconstructed from the center slice in half the time it takes to obtain data of the same statistical accuracy from the outer slices. By
65   offsetting the detector rings by half the angular

separation between detectors, it is seen that data is collected from the same number of points as would be collected from a single ring in its normal and rotated positions. Thus the imaged area is
70   sampled finely enough to allow an image to be reconstructed from a central slice without having to rotate the ring at all. This means in practice that images can be reconstructed from the central slice at a rate determined only by the amount of isotope
75   adminstered to the patient, the statistical accuracy required in the final image, and the transfer of raw data to more permanent storage (magnetic disc).

Referring now to Fig. 3, there is shown a sectional side elevation of adjacent portions of
80   two rings of detectors, U1 — U64 in one ring and L1 — L64 in the other ring.

The detector rings are separated by a thin tungsten septum 230 which prevents penetration of unwanted radiation from one detector ring to
85   the other. The adjacent detectors in the same row are also separated by respective tungsten septa 306, 308. The device also includes a pair of collimators in the form of annular end plates for preventing radiation outside the slice being
90   viewed to penetrate into one of the detectors. This is shown best in Fig. 2. An outer collimator comprises a first pair of annular lead plates 210, 212 which are displaceable towards and away from each other on rotatable screw-threaded rods
95   219 driven by a motor and belt drive (not shown) and toothed wheels 218. An inner collimator comprises a second pair of annular lead plates 223, 225 which are displaceable towards and away from each other on rotatable screw-
100   threaded rods 224 driven by another motor and belt drive (not shown) and toothed wheels 220. Reference is directed to our concorrectly filed U.K. Application No   entitled "Positron annihilation imaging device having movable
105   collimator" for further description of the collimator arrangement.

Figs. 3 and 4 show the positrons of the two rings of detectors 302, 304 with respect to one another. The detectors 302, 304 are angularly
110   separated by half the angle subtended by an individual detector.

In another embodiment a number of detector rings greater than two could be employed in which the detector arrays were alternately staggered by
115   half the angular separation of the detectors so that for N-rings of detectors (2N-1) cross-sectional images can be obtained.

The following advantages are achieved with regard to this invention.
120   More detectors view the radiation emanating from the patient to increase the use of the isotope administered to the patient, then if only one detector array was employed.

Encoding and processing data from coincident
125   events involving detectors in adjacent slices allows the whole volume to be imaged eliminating blind spots which would occur if the cross-slice coincident events were not used.

Using separate detectors on each slide (as
130   opposed to long detectors and electronically

encoding the location of the event within a detector) allows the count rate capability to be increased with no loss in efficiency. This allows the device to operate over a wider range in count
5 rate.

By rotating the detectors in adjacent rings by one-half the angular separation of the detectors, the imaging time in an even numbered slice can be improved by a factor of two. This is so since (a)
10 twice the number of detector pairs view the centre slice and (b) it is normally necessary to rotate the detector array by half the angular detector separation to achieve the desired spatial resolution and sampling of the imaged plane.
15 Other embodiments falling within the scope of the appended claims will occur to those skilled in the art.

CLAIMS
1. A positron annihilation imaging device
20 comprising a first array of N detectors equally spaced around a first circle and disposed in a first common plane; and a second array of N detectors equally spaced around a second circle and disposed in a second common plane, the first and
25 second circles being coaxial, said first and second planes being parallel and spaced apart, the detectors in the first array being angularly offset with respect to said detectors in the second array by an angle substantially equal to 360/2N°
30 subtended at the said common axis.
2. A positron annihilation imaging device as claimed in claim 1, further including at least one additional array of N detectors equally spaced around and coaxially circularly disposed in one or
35 more planes spaced from said common planes, the detectors in said additional array, being offset by 360/2N° relative to the detectors in an adjacent one of said first or second arrays of detectors.
40 3. A positron annihilation imaging device substantially as hereinbefore described with reference to the accompanying drawings.